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Patellofemoral instability in trochleodysplastic knee joints and the quantitative influence of simulated trochleoplasty - A finite element simulation

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Abstract: BACKGROUND Patellofemoral instability is a debilitating condition mainly affecting young patients and has been correlated with trochlear dysplasia. It can occur when the patella is insufficiently guided through its range of motion. Currently, there is no literature describing patellofemoral stability in trochleodysplastic knees and the effect of isolated trochleoplasty on patellofemoral stability. METHODS The effect of isolated trochleoplasty in trochleodysplastic knees of patients with symptomatic patellofemoral instability was investigated using a quasi-static finite element model. MRI data of five healthy knees were segmented, meshed and a finite element analysis was performed in order to validate the model. A second validation was performed by comparing simulated patellofemoral kinematics to in-vivo values obtained from upright- weight bearing CT scans. Subsequently, five trochleodysplastic knees were modelled before and after simulated trochleoplasty. The force necessary to dislocate the patella by 10 mm and to fully dislocate the patella was calculated in various knee flexion angles between 0 and 45°. FINDINGS The developed models successfully predicted outcome values within the range of reference values from literature. Lateral stability was significantly lower in trochleodysplastic knees compared to healthy knees. Trochleoplasty was determined to significantly increase the force necessary to dislocate the patella in trochleodysplastic knees to comparable values as in healthy knees. INTERPRETATION This is the first study to investigate lateral patellofemoral stability in patients with symptomatic patellofemoral instability and dysplasia of the trochlear groove. We confirm that patellofemoral stability is significantly lower in trochleodysplastic knees than in healthy knees. Trochleoplasty increases patellofemoral stability to levels similar to healthy.

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Patellofemoral instability in trochleodysplastic knee joints and the quantitative influence of simulated trochleoplasty – A finite element simulation☆

Dominik Kaiser[a,*], Linus Trummler[b], Tobias Götschi[a,b], Felix W.A. Waibel[a], Jess G. Snedeker[a,b], Sandro F. Fucentese[a]

[a] Department of Orthopedics, Balgrist University Hospital, Zurich, Switzerland
[b] Institute for Biomechanics, ETH Zurich, Zurich, Switzerland

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ABSTRACT

Background: Patellofemoral instability is a debilitating condition mainly affecting young patients and has been correlated with trochlear dysplasia. It can occur when the patella is insufficiently guided through its range of motion. Currently, there is no literature describing patellofemoral stability in trochleodysplastic knees and the effect of isolated trochleoplasty on patellofemoral stability.

Methods: The effect of isolated trochleoplasty in trochleodysplastic knees of patients with symptomatic patellofemoral instability was investigated using a quasi-static finite element model. MRI data of five healthy knees were segmented, meshed and a finite element analysis was performed in order to validate the model. A second validation was performed by comparing simulated patellofemoral kinematics to in-vivo values obtained from upright- weight bearing CT scans. Subsequently, five trochleodysplastic knees were modelled before and after simulated trochleoplasty. The force necessary to dislocate the patella by 10 mm and to fully dislocate the patella was calculated in various knee flexion angles between 0 and 45°.

Findings: The developed models successfully predicted outcome values within the range of reference values from literature. Lateral stability was significantly lower in trochleodysplastic knees compared to healthy knees. Trochleoplasty was determined to significantly increase the force necessary to dislocate the patella in trochleodysplastic knees to comparable values as in healthy knees.

Interpretation: This is the first study to investigate lateral patellofemoral stability in patients with symptomatic patellofemoral instability and dysplasia of the trochlear groove. We confirm that patellofemoral stability is significantly lower in trochleodysplastic knees than in healthy knees. Trochleoplasty increases patellofemoral stability to levels similar to healthy.

1. Introduction

Static (bony) stabilizers (Dejour et al., 1994; Malghem and Maldague, 1989; Senavongse and Amis, 2005), passive stabilizers (medial ligamentous complex) (Fithian et al., 2001; Hautamaa et al., 1998; Teitge et al., 1996) and active stabilizers (quadriceps muscle, especially vastus medialis obliquus) guide the patella through its range of motion. Static and passive stabilizers have a greater influence in early flexion compared to active stabilizers (Brossmann et al., 1994; Drew, 1908; Fithian et al., 1995).

Anatomical factors play a central role in patellar instability (Bollier and Fulkerson, 2011; Dejour et al., 1994) including trochlear and patellar dysplasia, an increased tibial tuberosity- trochlear groove (TT-TG) distance, a patella alta, increased Q-angle, a long patellar tendon, axial plane alignment and a valgus leg axis (Atkin et al., 2000; Dejour et al., 1994; Dejour and Locatelli, 2001; Neyret et al., 2002; Nomura et al., 2000; Panni et al., 2011; Wiberg, 1941). The medial patellofemoral ligament (MPFL) additionally stabilizes the patella in extension

Abbreviations: FE, finite element; LPFL, lateral patellofemoral ligament; MPFL, medial patellofemoral ligament; MPML, medial patellomeniscal ligament.

☆ Investigation performed at Balgrist University Hospital, Zurich, Switzerland

* Corresponding author at: Universitätsklinik Balgrist, Forchstrasse 340, Zürich 8008, Switzerland.
E-mail address: dominik.kaiser@balgrist.ch (D. Kaiser).

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and early flexion (Nomura et al., 2000). A disruption of the MPFL has been reported in more than 90% of first-time patella dislocations (Desio et al., 1998; Hautamaa et al., 1998; Nomura et al., 2000). The goal of surgical interventions to restore patellar stability may consist of (a) reconstruction of the medial patellofemoral ligament (MPFL) (b) realignment of the extensor mechanism by reducing the TT-TG, correcting the valgus leg axis, correcting the axial plane alignment and or lateral lengthening (c) reconstruction of the trochlear groove as well as (d) correcting patellar height.

The clinical results of surgical treatment of patellar instability are promising both in short- and in long-term (Arendt, 2009; Bereiter and Gautier, 1994; Chassaing and Tremoulet, 2005; Nomura et al., 2007; Schöttle et al., 2009; Von Knoch et al., 2006), although premature patellofemoral osteoarthritis remains a major concern in these patients (Dejour and Allain, 2004; Grelsamer et al., 2008). Literature on the quantitative assessment of patellar stability is scarce. Cadaver studies quantifying patellar stability has only been performed in healthy knees (Amis et al., 2008; Conlan et al., 1993; Desio et al., 1998; Farahmand et al., 1998; Farahmand et al., 2004; Marumoto et al., 1995; Nomura et al., 2000; Senavongse et al., 2003). Such information is critical to understand the functional implications of the performed procedures, and to provide guidance on the minimal amount of anatomical correction that is necessary to restore stability.

As it is practically impossible to obtain dysplastic knees for post mortem experimental investigation, we relied on computational models to investigate patellofemoral biomechanics in patients with trochlear dysplasia. Computational models allow for a wide range of parametric variation of input parameters, enabling identification of the key factors related to joint dysfunction and potential for surgical restoration.

The purpose of this study was to quantitatively assess the force necessary to lateralize the patella by 10 mm and fully dislocate the patella in healthy knees, in knees of patients with symptomatic patellar instability and a dysplastic trochlea and after simulated surgical trochleoplasty.

We employ the model to test the hypotheses (a) The force necessary to lateralize the patella 10 mm and fully dislocate the patella in trochleodysplastic knees is significantly lower than in healthy knees (b) Trochleoplasty increases the force necessary to lateralize the patella 10 mm and fully dislocate the patella.

2. Methods

2.1. Ethical approval

Approval of the ethics committee (BASEC Nr. 2018-01447) and informed consent of all patients was obtained.

2.2. MRI assessment and model validation

MR images of the tested knees were acquired with a sagittal high-resolution isotropic 3D PD SPACE sequence. A voxel size of 0.7 × 0.7 × 0.7mm, and a field of view of 152 mm x170mm, was imaged using a 3 Tesla Siemens Skyra MR Scanner (Siemens, Munich, Germany) with a 15 channel transmit/receive coil. Echo time was 9.4 ms and repetition time was 700 ms.

MR image data of five healthy adult knee joints were selected to have a comparable cohort to the six fresh frozen cadaver knees used in the literature (Amis et al., 2008; Farahmand et al., 2004). The patients had no history of patellar instability and no anatomical risk factors. None of the patients had a trochlear or patellar dysplasia, TT-TG was <10 mm, patellar height was normal (Caton-Deschamps Index 0.75–0.89) and lateral trochlear inclination was >16°. The MRI DICOM data was segmented semi-automatically using the freeware 3D-Slicer (Version 4.10.1) and improvement of the models mesh was performed according to Kumara and Pietroni (Kumara, 2011; Pietroni et al., 2009) using Meshlab 1.3.3 (Computing Lab-ISTI-CNR). Simulations of different flexion-angles of the knee were performed according to Kurosawa et al. (Kurosawa et al., 1985). They provided quadratic equations for the rotation and translation of the femoral origin in relation to the tibial origin. Hence, we calculated the tibiofemoral positions for every flexion-angle using these equations. The transformations are hereby specified in a landmark-based coordinate system with its origin located in the center of the two femoral condyles aligned with the long axis of the femur. Patellar kinematics were derived from the prescribed tibiofemoral kinematics in a static structural finite element (FE) simulation (ANSYS® Academic Research, 19.2). The final position of the patella is a result of the forces of the quadriceps, the tension in the ligaments and the contact between patella and femur. The validity of this approach has been demonstrated by Baldwin et al. (Baldwin et al., 2009).

In an effort to provide best possible comparability for the validation (Viceconti et al., 2005) of the model the quadriceps was loaded with a total of 175 N divided into the rectus femoris + vastus intermedius (35% with direction 0° mediolateral and 0° anterior), vastus medialis obliquus (9% with 47° medial and 44° posterior), vastus medialis lateralis (14% with 15° medial 0° anterior), vastus lateralis obliquus (9% with 35° lateral and 33° posterior) and vastus lateralis longus (33% with 14° lateral and 0° anterior) (Amis et al., 2008) (Table 1). Ligamentous patellar stabilizers were modelled as tension-only springs. The MPFL and the lateral retinaculum were represented by four tension-only springs with a total stiffness of 12 N/mm and 2 N/mm, respectively, the medial patellomeniscal ligament (MPML) was represented by three tension-only springs and had a combined linear stiffness of 5 N/mm according to Elias et al. (Elias and Cosgarea, 2006). They were modelled with an elastic and linear behavior without a predetermined maximum. Pretension was set to 0 in the starting position before applying the quadriceps load. Attachment sites were obtained directly from the MR image (quadriceps tendon, patellar tendon) or from the literature (MPFL, MPML, lateral retinaculum) (Elias and Cosgarea, 2006). The patellar tendon was modelled with five tension-only springs, while mechanical properties of the patellar tendon were derived from the literature (Hansen et al., 2006). In order to maximize comparability to the literature (Amis et al., 2008), where all muscles apart from the quadriceps, and all soft tissue was excised apart from the capsule and the retinacular structures, no further anatomical structures were included in our model. Contact behavior between the femur and the patella was modelled as frictional with a friction coefficient of 0.02 (Oungoulian et al., 2015; Shah et al., 2015). Cartilage was modelled using rigid surface elements and deformation was considered by formulating contact behavior with a penetration penalty. Bones were treated as rigid structures (Donahue et al., 2002) and the femur was fixed in space to prevent rigid body motion (Fig. 1). The force necessary to lateralize the patella 10 mm and the peak force to fully dislocate the patella was then calculated in healthy knees in the flexion angles 0°, 10°, 20°, 30°, 45°. Patellar position was visualized to verify a possible dislocation. We assumed that the patella was fully dislocated if more than half of the patella was dislocated over the lateral ridge of the trochlea (Fig. 2). These results were compared to the literature (Amis et al., 2008) to validate our model.

Table 1
Comparison of model parameters applied in the current study compared to the parameters used in the literature.

	Current study	Literature
Model type	Computational model	In vitro cadaver study
Knee pathology	None	none
Soft tissue	Not included	removed
Quadriceps force	175 N (total load)	175 N (total load)
Force vector	Literature	Literature
Friction coefficient cartilage	0.02	–
Patellar tendon	Literature	–
MPFL, MPML, Lat ret.	Literature	–
Femorotibial Kinematics	Literature	–

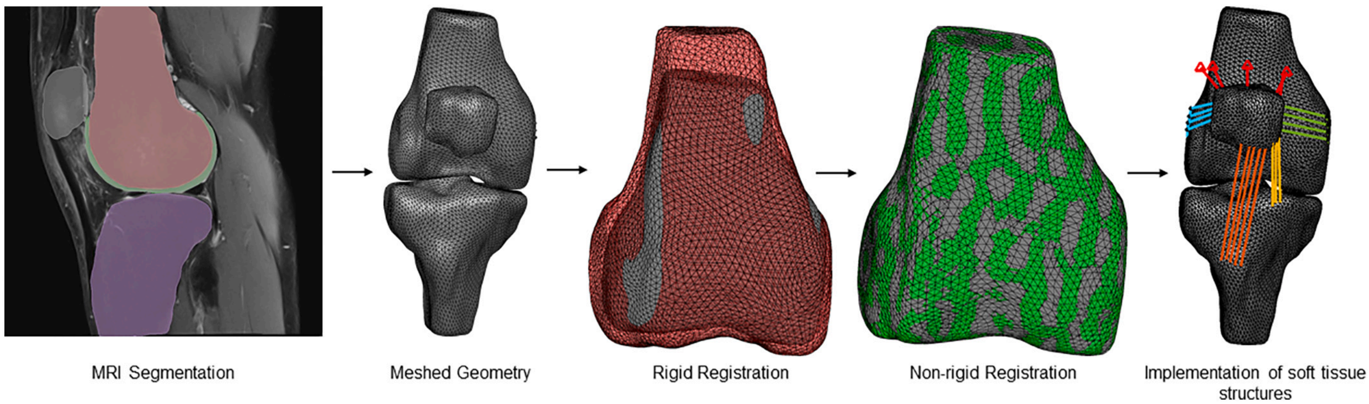

Fig. 1. Workflow for the model creation of a new knee. Rigid and non-rigid registration is performed separately on femur, tibia and patella. In the rigid and non-rigid registration (here at the example of the femur), the red femur represents the template, the gray femur is the new femur and the green femur illustrates the adapted geometry of the new femur. The quadriceps muscle (red), MPFL (green), MPML (yellow), lateral retinaculum (blue) and patellar tendon (orange) are visualized. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

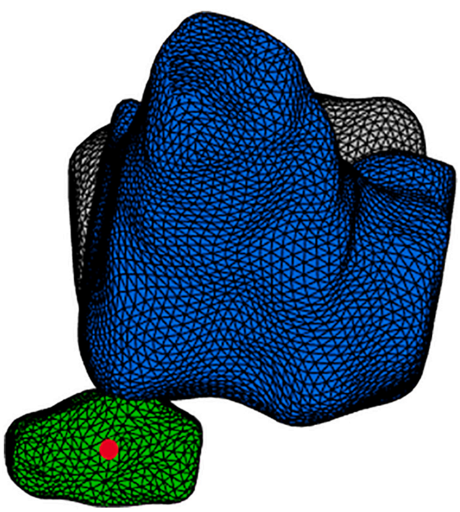

Fig. 2. Axial view of our FE model (femur in blue, tibia in gray, patella in green, center of the patella as a red dot). The patella was assumed to be fully dislocated when the center exceeded the lateral trochlea. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

A second validation was performed to increase validity of our FE model further. Patellar position at 0°, 30° and 60° from five upright weight- bearing CT scans of healthy knees were extracted and compared with our simulated patellar positions of the same knees (Hirschmann et al., 2017). The starting position of the patella was hereby determined from the 0° CT images. After simulated 30° and 60° knee flexion the predicted mediolateral position of the patella was compared to its respective position in the CT images (Smith et al., 2009).

2.3. *Investigation of the trochleodysplastic knees*

After successful validation of our model the MR image data of five patients with symptomatic patellar instability who had undergone surgical stabilization at our institution were segmented from the MRI performed prior to the surgery. Demographic data and anatomical properties of this second cohort are summarized in Table 2. The starting position of the patella was obtained from the MRI. The scarred MPFL was given the same stiffness as the native MPFL, as we did not find any values in the literature, whereas the length was obtained indirectly by segmenting the patellar position from the MRI image data. The

Table 2
Demographic data and anatomical properties of the trochleodysplastic knees.

Side	Left	Right	Left	Left	Right
Sex	F	M	M	F	F
Age [a]	23	20	25	15	20
Height [cm]	175	173	191	173	176
Weight [kg]	63	63	109	82	60
Trochlear dysplasia[a]	B	B	B	B	B
Patellar dysplasia[b]	III	III	III	III	III
Leg axis [°]	1 valgus	2 valgus	1 varus	2 valgus	2 valgus
Lat. Trochlear inclination [°]	7	5	7	4	10
Femoral Torsion [°]	26	8	7	7	n/a
TT-TG [mm]	19	18	21	12	7
Patellar height (Caton Deschamps Index)	1.25	1.06	1.19	1.33	1.24

[a] According to Déjour.
[b] According to Wiberg.

mechanical properties of the ligaments were identical as in the above-mentioned model. The force necessary to lateralize the patella 10 mm and the peak force to fully dislocate the patella was then calculated in the same angles. Patellar position was visualized to confirm patellar dislocation.

2.4. *Trochleoplasty*

Trochleoplasty in a modified Bereiter technique (Bereiter and Gautier, 1994) was performed freehand using modeling software (Blender 2.72, Blender Foundation, Netherlands) and confirmed by the senior author. The patella was then brought closer to the femur again so that normal contact was reestablished; this led to a medialization of the patellar starting point of approx. 7 mm. Ligament length was kept the same. The force necessary to lateralize the patella 10 mm and the peak force to fully dislocate the patella was then calculated in the flexion angles 0°, 10°, 20°, 30°, 45°. Patellar position was visualized to confirm patellar dislocation.

2.5. *Statistical analysis*

Agreement between simulated and predicted mediolateral patella positions was assessed using Pearson correlation. Statistical analysis was performed using a two-tailed non-parametric independent test (Mann Whitney *U* test) for the comparison of the force values. Differences were considered to be statistically significant for *p*-values <0.05. Results are reported as mean, standard deviation and associated p-values if not stated otherwise.

3. Results

3.1. Model validation with healthy knees

First validation: The calculated forces for the different flexion angles for the validation of our computational model were in good agreement with values reported in literature (Amis et al., 2008) being well within reported standard deviations (Fig. 3).

Second validation: A Pearson correlation test was performed with a satisfactory overall agreement with a coefficient of determination of $R^2 = 0.687$ (Fig. 4). Visual inspection revealed that the model predicts a slightly lateralized path from 0° to 60° flexion in comparison to the same knee in-vivo (Fig. 4).

3.2. Trochleodysplastic knees before and after trochleoplasty

The demographic and anatomical parameters of the segmented trochleodysplastic knees are summarized in Table 2. All knee joints had a trochlear dysplasia type B (Dejour et al., 1994), patellar dysplasia type III (Wiberg, 1941) and only a minimal deviation of 2° from a straight leg axis. The patellar height ranged from 1.06–1.33 as measured by the Caton-Deschamps-Index in all knees, the TT-TG ranges from 7 to 21 mm as measured on MRI and the lateral trochlear inclination ranged from 4 to 10°. The results of the FE calculations for all three groups are summarized in Tables 3, 4 and Fig. 5. In the trochleodysplastic knees a dislocation of 12, 13 and 14 mm was necessary to fully dislocate the patella at 45° flexion leading to a greater average peak force (Tables 3, 4). This did not influence the statistical difference.

4. Discussion

This is the first study to provide objective quantitative values for the lateral stability of the patella in knees with a trochlear dysplasia type B (Dejour et al., 1998) of patients with a symptomatic patellar instability after failed conservative treatment. The force necessary to lateralize the patella was calculated in the relevant first 45° of flexion. This is also the first study to calculate the clinically more relevant force to completely dislocate the patella. Our study confirms that the force necessary to lateralize the patella 10 mm and fully dislocate the patella is significantly lower in patients suffering from patellar instability and a trochlear dysplasia type B. In this specific subgroup isolated trochleoplasty may increase patellar stability significantly to a level comparable to healthy knees from 10° to 45° of flexion. We explain this result by the fact that the patella engages in the trochlear groove in early flexion but not in full extension (Tables 3, 4).

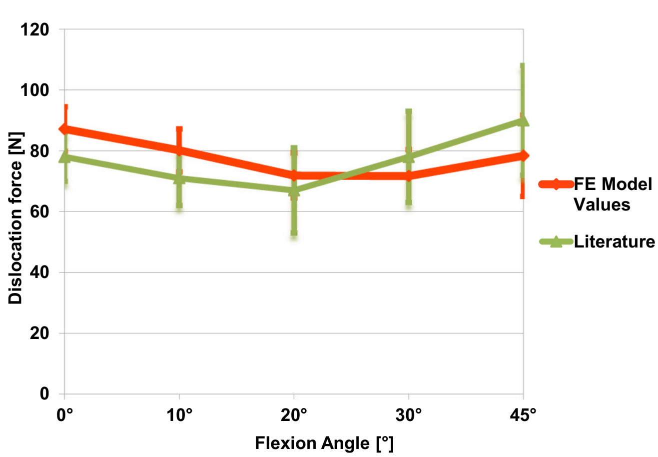

Fig. 3. Graphic model validation between the reported mean values of the force necessary to laterally dislocate the patella by 10 mm (Amis et al., 2008) compared to the mean values of the force necessary to laterally dislocate the patella by 10 mm in our model.

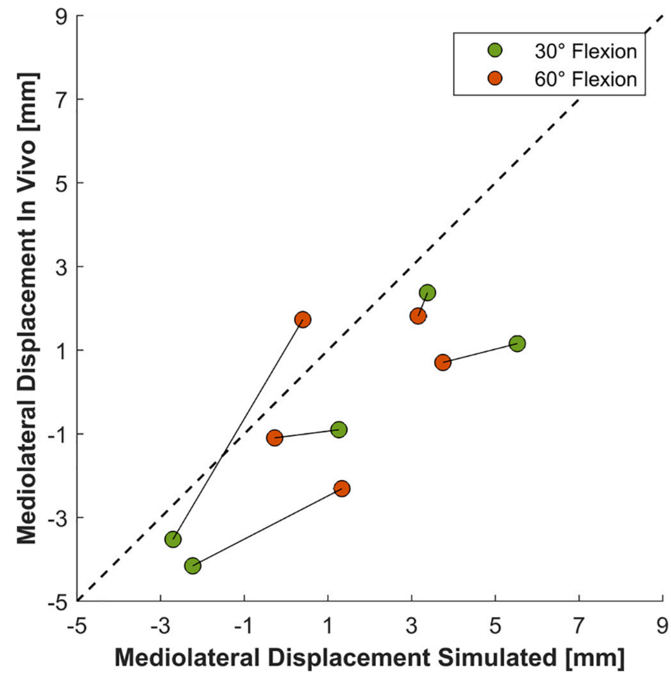

Fig. 4. Linear regression plot with simulated and in-vivo mediolateral positions. While green colored data points represent knees at 30° flexion, orange data points represent knees at 60° flexion. Knees of the same patient are connected through a black line. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

An important asset of the current investigation is the fact that all included knees stem from patients with recurring patellar instability where conservative treatment failed. In an effort to minimize confounding, only knees with the same grade of trochlear and patellar dysplasia as well as with a relatively straight leg axis were included (Table 2). In this way remaining factors influencing patellar stability such as patellar height, its position and geometry (Wiberg, 1941), the length of the patellar tendon (Neyret et al., 2002) and the tibial tuberosity could be obtained directly from the MRI and included in the model. As patellar stability is multifactorial we are convinced that this approach reproduces reality in the best possible way and more complete than biomechanical experiments on healthy knee joints. A possibility to individualize the model even further would be to obtain the relative cross-section area of the muscles directly from the MRI data. However, in order to compare and validate our model to the literature (Amis et al., 2008) it was necessary to load the model identically. To ensure comparability between the groups the muscle load was not altered. As the aim of this study is to investigate the isolated effect of trochleoplasty, we have deliberately changed as few parameters in our FE model as possible. The new starting point of the patella was approx. 7 mm medial after trochleoplasty. This leads to a relaxation of the medial structures (MPFL) and to a pretension of the lateral structures (lateral retinaculum). As the MPFL is a stiffer structure than the lateral retinaculum in our model this may have led to an underestimation of the stabilizing effect of the trochleoplasty in both calculations, as the stabilizing effect of the MPFL was reduced. However, an arbitrary change in the length or the tension of these structures would have had an additional possibly diluting influence on our calculations of the isolated effect of trochleoplasty. While it is of great importance for the surgeon to understand the effect of isolated trochleoplasty on patellar stability; trochleoplasty is not performed isolated but when necessary in addition to a

Table 3
Summary of the calculated force values necessary to lateralize the patella by 10 mm in all three groups in different flexion angles. (Significance level *P < 0.05)*.

	Healthy knees			Dysplastic trochlea			Trochleoplasty		
Flexion angle [°]	Force 10 mm [N]	SD	Vs dysplastic *P*=	Force 10 mm [N]	SD	Vs trochleoplasty *P* =	Force 10 mm [N]	SD	Vs. healthy *P* =
0	87	7	0.012	66	4	n.s.	65	10	0.037
10	80	7	0.021	55	23	0.037	80	13	n.s.
20	71	7	0.021	47	11	0.037	70	28	n.s.
30	70	8	0.012	37	8	0.012	87	25	n.s.
45	68	32	0.012	22	10	0.012	85	30	n.s.

Table 4
Summary of the calculated force values necessary to fully dislocate the patella in all three groups in different flexion angles. (Significance level *P < 0.05)*.

	Healthy knees			Dysplastic trochlea			Trochleoplasty		
Flexion angle [°]	Peak Force [N]	SD	Vs dysplastic *P*=	Peak Force [N]	SD	Vs trochleoplasty *P*=	Peak Force [N]	SD	Vs. healthy *P*=
0	89	7	0.012	67	3	n.s.	69	10	0.021
10	80	7	0.021	61	12	0.037	80	13	n.s.
20	72	7	0.021	52	6	0.012	82	18	n.s.
30	74	12	0.012	38	8	0.012	88	23	n.s.
45	80	17	0.012	35	5	0.012	89	34	n.s.

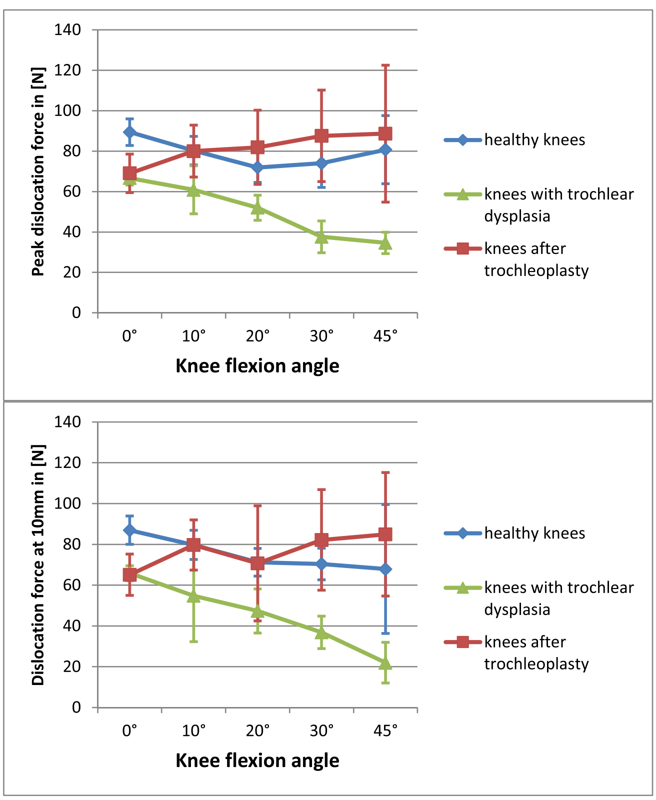

Fig. 5. Peak lateral dislocation force (above) and dislocation force necessary to lateralize patella by 10 mm (below) including standard deviation at different knee flexion angles.

reconstruction of the MPFL.

With this FE model we will continue to investigate the influence of additional patellar stabilizers such as reconstruction of the isolated medial patellofemoral ligament as well as MPFL-reconstruction with simultaneous trochleoplasty, medializing osteotomy of the tibial tuberosity and the effect of targeted strengthening of the vastus medialis muscle. Deepening the understanding of the effects of various parameters and their interaction in relation to patellofemoral joint stability may enable improved decision-making regarding case-specific intervention. Further, the available analysis pipeline may prove useful in optimizing trochlear shape during trochleoplasty restoring stability in the full range of motion without overcorrection in 30–45° of flexion while keeping the patellofemoral pressure on the cartilage as low as possible.

Despite FE modeling being the best way to investigate patellar stability in trochleodysplastic knee joints, there are numerous limitations to this simulation study.

Computational models inherently simplify reality. The primary concern of computational modeling is the relevance of the output and relationship to the modelled input parameters. Model validation with available in-vitro experiments (Amis et al., 2008) was performed to minimize this limitation. Additional model validation with direct comparison of in-vivo patellar tracking with our model was performed. The mechanical characteristics of ruptured and scarred MPFLs are difficult to determine experimentally and are not available in the literature. We assumed that the scarred MPFL would have a similar stiffness compared to the native MPFL, albeit with an elongated length which was segmented directly from the MRI. As it was our goal to investigate the isolated effect of trochleoplasty in dysplastic knee joints we did not vary the stiffness and the slack of the MPFL, as this would have diluted our results. We acknowledge that we have most likely overestimated the effect of the MPFL by assuming that the scarred MPFL is comparable to a healthy MPFL and that patellar stability may be even lower in reality. As the MPFL has been described to be isometric in extension up to 100 degrees we believe our approach of obtaining the MPFL length from the MRI image is justified (Nomura et al., 2005; Smirk and Morris, 2003). Increasing the number of cases in the model validation as well as in the trochleodysplastic group will most likely not change the principal findings we have made, but would have increased confidence in its generalizability. The results of the comparison of the dysplastic knee group after trochleoplasty and the healthy knees should be interpreted with caution as we compare different groups of knees. We are aware that most patellar dislocations occur under greater quadriceps loading conditions and will investigate the influence of quadriceps loading on stability in further simulations.

5. Conclusion

This is the first study to demonstrate that lateral patellar stability is significantly lower in trochleodysplastic knees compared to healthy knees. Isolated trochleoplasty can significantly increase patellar stability from 10° to 45° flexion compared to trochleodysplastic knees to levels comparable to healthy knees. This FE model study underlines the importance of trochleoplasty as part of a treatment concept for patellar instability in patients with a trochlear dysplasia type B.

Declaration of Competing Interest

None

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